includes a solenoid 204 which can be either a bifilar winding (as shown in Figure 4) or a simple winding (as shown in Figure 6). A plurality of mechanical springs 206 regulate a null level and a trigger level to control a metal piston on/off condition. Mounted within piston assembly 209 is a plurality of pistons 210. When no net magnetic field disturbances except original magnet field created by the main and shielding coils present in solenoid 204, mechanical springs 206 are at pre-set null level, and metal pistons 210 do not contact a stator, and hence, no current goes through the resistive quench heater(s). When electrical current reaches a pre-set level (e.g., 2 amps) in solenoid 204 by the environment disturbances, the electromagnetic force on pistons 210 pulls one of the pistons 210 toward the stator, and the quench heater circuit engages, and cause the superconducting wires (CC'D'D and/or D'D"E'E) to quench. When the current drops to zero after quench, piston 210 returns to its null position, and the quench circuit is disengaged. In one embodiment, pistons 210 are positioned opposing each other such that current flow in either direction CC'D'D or DD'C'C causes one of pistons 210 to move toward a center of assembly 209 to complete the circuit between power supply 201 and heater 202. In an alternative embodiment, only a single piston 210 is used.

Remarks

Please enter the foregoing preliminary amendment prior to examination of the present application. Submitted herewith are marked up Paragraphs in accordance with 37 C.F.R. 1.121(b)(1)(iii). Also submitted herewith is a request for drawing change correcting a mis-numbering of power supply 201.

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PATENT

IN THE UNITED STATES PATENT AND TRADEMARK OFFICE

Applicant: Weijun Shen et al.	:	Art Unit: Not yet assigned
Serial No.:	:	Examiner: Not yet assigned
Filed:	:	
For: METHODS AND APPARATUS FOR IMAGING SYSTEMS	:	

SUBMISSION OF MARKED UP PARAGRAPHS

Hon. Commissioner for Patents
P.O. Box 1450
Alexandria, VA 22313-1450

Submitted hereby are marked up Paragraphs in accordance with 37 C.F.R. 1.121(b)(1)(iii).

Please delete paragraph 1, and replace therefore with the following replacement paragraph:

This invention relates to a Magnetic Resonance Imaging (MRI) system. More particularly, this invention relates to superconducting coils used in MRI systems for correcting central magnetic field temporal shift[ ], and for shielding external magnetic disturbances from large electromagnetic fields.

Please delete paragraph 2, and replace therefore with the following replacement paragraph:

A highly uniform magnetic field is useful for nuclear resonance image (MRI) and nuclear magnetic resonance (NMR) systems as medical devices or chemical/biological devices. Most popular systems currently available worldwide use a superconducting magnet system which creates a highly uniform field in a pre-determined space (imaging volume). A

superconducting magnet system usually uses multiple superconducting coils (main coil system) to achieve a desired high uniform magnetic field in the imaging volume. More advanced superconducting MRI and NMR magnet systems also use[s] an active shielding technique[-] which adds a second set of multiple coils (shielding coil system) which creates a reverse direction magnetic field to reduce the fringe magnetic field and to achieve a significant reduction of the external magnetic field in the surrounding space of the magnet system. Depending on the design, the main coils system and shielding coils system can use a single circuit running the same electrical current, or two individual circuits running either the same current or two different currents. From the laws of physics, one knows that for a single superconductive closed loop, the total magnetic flux inside of the loop does not change. However, a multi-coil system, especially for an actively shielded MRI magnet system with main coils and shielding coils connected in series, the situation is a little different.

Please delete paragraph 3, and replace therefore with the following replacement paragraph:

Due to the environment disturbances, such as a train[,]and/or other moving vehicles, rotating machinery, elevators etc, in the surrounding area, the magnetic field of the system will have a corresponding temporal magnetic flux change. Practically, all magnet systems are subject to such temporal field instability ranging from ppm (parts per million) to ppb (parts per billion). But for actively shielded magnet systems, this change is more severe. For good image quality, the temporal field variation of a typical MRI should normally be less than 0.05 to 0.1 ppm/hour. The stability of the magnet center field is, however, highly affected by the environment disturbances, especially for those actively shielded magnets. The magnitude of the field fluctuations depends on both the size of the object and the distance away from the magnet system. For example, a typical elevator[s] 20 feet away from the magnet can cause a field fluctuation of about 0.01 Gauss or 1.0E-6 Tesla, a subway can also cause a 0.1 Gauss field fluctuation.

Please delete paragraph 5, and replace therefore with the following replacement paragraph:

In order to minimize such effect caused by environment changes and other disturbances, the electrical currents changing in both main coils and shielding coils should be controlled or limited to some prescribed acceptable level such that the environment disturbance is compensated and the center magnetic field remain constant and uniform[ ]. One structure and method has been described in US Patent No. 4926289 for such purpose by using a single filament or a few filaments of superconducting wire for the purpose of having low critical current. However, it would be desirable to provide methods and apparatus which are not constrained to filament(s) with low critical current.

Please delete paragraph 10, and replace therefore with the following replacement paragraph:

Figure 2 is a schematic diagram of a conventional circuitry of superconducting current limiter for a superconducting MRI, NMR magnet system.

Please delete paragraph 17, and replace therefore with the following replacement paragraph:

Herein described are methods and apparatus which utilize a current limiter for active shielding of a superconducting magnet system used in MRI and NMR magnetic field generators. More specifically, in one embodiment, a detection system is provided for an active shielding of superconducting magnet systems which use a single electrical current as explained in greater detail below. In another embodiment, a detection system is provided for an active shielding of a multiple electrical circuits superconducting magnet system[s] as also explained in greater detail below. The herein described methods and apparatus use a combination of a detection mechanism and a controlled triggering level to limit the electrical current induced by environment disturbances.

Please delete paragraph 19, and replace therefore with the following replacement paragraph:

Figure 1 is a block diagram of an embodiment of a magnetic resonance imaging (MRI) system 10 in which the herein described systems and methods are implemented. MRI 10 includes an operator console 12 which includes a keyboard and control panel 14 and a display 16. Operator console 12 communicates through a link 18 with a separate computer system 20 thereby enabling an operator to control the production and display of images on screen 16. Computer system 20 includes a plurality of modules 22 which communicate with each other through a backplane. In the exemplary embodiment, modules 22 include an image processor module 24, a CPU module 26 and a memory module 28, also referred to herein as a frame buffer for storing image data arrays. Computer system 20 is linked to a disk storage 30 and a tape drive 32 to facilitate storing image data and programs. Computer system 20 [is] communicates with a separate system control 34 through a high speed serial link 36.

Please delete paragraph 20, and replace therefore with the following replacement paragraph:

System control 34 includes a plurality of modules 38 electrically coupled using a backplane (not shown). In the exemplary embodiment, modules 38 include a CPU module 40 and a pulse generator module 42 that is electrically coupled to operator console 12 using a serial link 44. Link 44 facilitates transmitting and receiving commands between operator console 12 and system command 34 thereby allowing the operator to input a scan sequence that MRI system 10 is to perform. Pulse generator module 42 operates the system components to carry out the desired scan sequence, and generates data which <u>is</u> indicative of the timing, strength and shape of the RF pulses which are to be produced, and the timing of and length of a data acquisition window. Pulse generator module 42 is electrically coupled to a gradient amplifier system 46 and provides gradient amplifier system 46 with a signal indicative of the timing and shape of the gradient pulses to be produced during the scan. Pulse generator module 42 is also configured to receive patient data from a physiological acquisition controller 48. In the exemplary embodiment, physiological acquisition controller 48 is configured to receive inputs from a plurality of sensors indicative of a patient's physiological condition such as, but not limited to, ECG signals from electrodes attached to the patient. Pulse generator module 42 is electrically coupled to a scan room interface circuit

50 which is configured to receive signals from various sensors indicative of the patient condition and the magnet system. Scan room interface circuit 50 is also configured to transmit command signals such as, but not limited to, a command signal to move the patient to a desired position[,] with [to] a patient positioning system 52.

Please delete paragraph 23, and replace therefore with the following replacement paragraph:

The NMR signals received by RF coil 66 are digitized by transceiver module 70 and transferred to a memory module 78 in system control 34. When the scan is completed and an array of raw k-space data has been acquired in the memory module 78[.], [T]the raw k-space data is rearranged into separate k-space data arrays for each cardiac phase image to be reconstructed, and each of these arrays is input to an array processor 80 configured to Fourier transform the data into an array of image data. This image data is transmitted through serial link 36 to computer system 20 where it is stored in disk memory 30. In response to commands received from operator console 12, this image data may be archived on tape drive 32, or it may be further processed by image processor 24 and transmitted to operator console 12 and presented on display 16.

Please delete paragraph 26, and replace therefore with the following replacement paragraph:

Figure 4 illustrates a one coil detection system 150 in which MRI system 10 includes a cryogenic temperature cryostat 152 in which a main coil 154, a shielding coil 156, a quench protection system 158, and a superconducting persistent switch 160 are positioned. A power supply 161 is typically positioned outside cryostat 152. System 150 also includes an environmental fluctuation circuit 162. In an exemplary embodiment, main coil 154 and shield coil 156 are wired in series receiving the same current, and environmental fluctuation circuit 162 is wired in parallel to one of main coil 154 and shielding coil 156. As illustrated in Figure 4, environmental fluctuation circuit 162 is wired in parallel to main coil 154. When electrical current I_a and I_b are not equal due to outside electromagnetic disturbances, the differential current of main coils I_a and shielding coils I_b flows through superconducting

circuit CC'D'D, thus with the aid of detection and controlling scheme identical or similar to that illustrated in Figure 6, a differential current I_c is detected, limited, and/or controlled. Although Figure 4 illustrates that superconducting wire is connected to main coil 154 at point C and D in Figure 4, the superconducting wire alternatively can be connected to shield coil 156 [with the same concept]<u>similarly</u>, or be connected to the points within the coil. For example, in Figure 4 points C and D are located at a plurality of edges of coil 154, points C and D may be located within coil 154 and coil 156 respectively (i.e., points C and/or D are located in a coiled section of coil(s) 154 and/or 156). The exact position of points C and D for example depends entirely on a particular magnet design and the requirements for environment disturbance compensation. Figure 5 through Figure 8 explain in additional detail how to detect these induced currents and how to control/eliminate these currents.

Please delete paragraph 27, and replace therefore with the following replacement paragraph:

Figure 5 is a detailed illustration of a detection circuit 170 having two parts, one part is connected to points C, D, and E of Figure 1, with two pieces of superconducting wire 176 and 178 wound on a single mandrel in bifilar fashion, the other part is a plurality of quench heaters 174 with a controlling switch 180 and a resistive quench heater power supply 172. A sensor 182 is positioned to sense electromagnetic fields. When the current either in CC'D'D circuit (I_c) or DD"E'E (I_d) or both starts to flow, and with the aid of detection sensor 182 (either mechanical or electronic as detailed below) and control switch K, quench heaters 174 are energized to heat the superconducting wires CC'D'D and DD"E'E and cause the superconducting wire to quench when current I_c and/or I_d reaches above a predetermined level (e.g., 2 amperes), and thus reduce the electrical currents I_c and I_d to zero, which forces electrical currents in main coil 122 I_a and shield coil 124 I_b to be the same. After sensor 182 detects zero current in I_c and/or in I_d, control switch 180 switches off the current in the quench heaters 174. Thus the electrical currents of main coil 122 and shield coil 124 are the same again. A similar [concept] <u>construction</u> is also shown in Figure 6 for one coil detection circuit 150 (shown in Figure 4).

Please delete paragraph 29, and replace therefore with the following replacement paragraph:

Figure 7 is a schematic of a mechanical sensor 200 for detection systems 118 and 150 (e.g., sensors 182 and 196), employed in some embodiments. A power source 20[0]1 is coupled to a quench heater 202 via wires 208 to a piston assembly 209. Mechanical sensor 200 includes a solenoid 204 which can be either a bifilar winding (as shown in Figure 4) or a simple winding (as shown in Figure 6). A plurality of mechanical springs 206 regulate a null level and a trigger level to control a metal piston on/off condition. Mounted within piston assembly 209 is a plurality of pistons 210. When no net magnetic field disturbances except original magnet field created by the main and shielding coils present in solenoid 204, mechanical springs 206 are at pre-set null level, and metal pistons 210 do not contact a stator, and hence, no current goes through the resistive quench heater(s). When electrical current reaches a pre-set level (e.g., 2 amps) in solenoid 204 by the environment disturbances, the electromagnetic force on pistons 210 pulls one of the pistons 210 toward the stator, and the quench heater circuit engages, and cause the superconducting wires (CC'D'D and/or D'D"E'E) to quench. When the current drops to zero after quench, piston 210 returns to its null position, and the quench circuit is disengaged. In one embodiment, pistons 210 are positioned opposing each other such that current flow in either direction CC'D'D or DD'C'C causes one of pistons 210 to move toward a center of assembly 209 to complete the circuit between power supply 20[0]1 and heater 202. In an alternative embodiment, only a single piston 210 is used.

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